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(71) Applicant: **PICKER INTERNATIONAL, INC.**
Highland Heights Ohio 44143 (US)

(72) Inventors:
- **Morich, Michael A.**
  Mentor, Ohio 44060 (US)
- **Petropoulos, Labros S.**
  Solon, Ohio 44139 (US)

(74) Representative: **Pope, Michael Bertram Wingate**
The General Electric Company, p.l.c.
GEC Patent Department
Waterhouse Lane
Chelmsford, Essex CM1 2QX (GB)

(54) Magnetic resonance imaging apparatus

(57) A magnetic resonance imaging apparatus includes superconducting magnets (10) which generate a temporally constant magnetic field through a bore (12). The bore, hence the superconducting magnets, have a length which is relatively short compared to its diameter, eg a length to diameter ratio of less than 1.5 to 1.0 and typically 1:1. A gradient coil assembly (30) of the apparatus is disposed around the bore for generating gradient magnetic fields across the bore. With such a relatively short bore magnet, in the region in which the gradient field coil is disposed, the main magnetic field suffers non-uniformities including radial magnetic field components. When current pulses are fed through the windings of a primary gradient coil (32) and a secondary gradient coil (34) of the gradient coil assembly the currents interact in an unbalanced manner with the non-uniformities and radial components of the temporally constant magnetic field, causing a net force in axial and/or transverse directions. Force correction windings (64, 72) are mounted adjacent the ends of the gradient coil, carrying current in an opposite direction to z and transverse gradient windings in order to produce a counterbalancing force such that the net and counterbalancing forces substantially cancel. In this manner, the gradient coil assembly is freed from the necessity of a substantial mechanical mountings and the vibration and distortion which large net forces can cause.

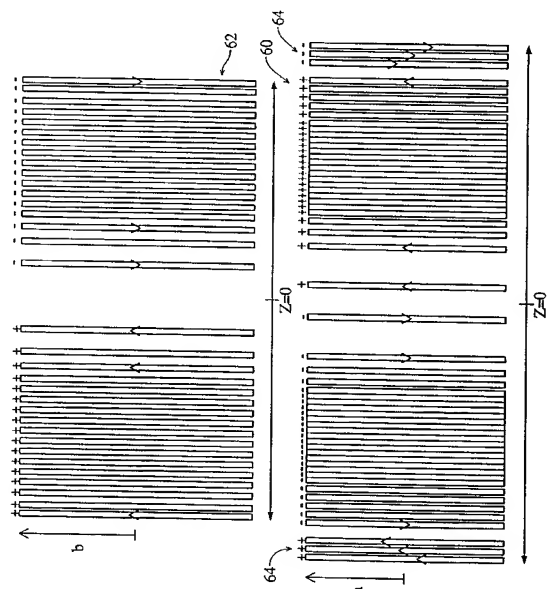

Fig.2B

Fig.2A

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Description

This invention relates to magnetic resonance imaging apparatus. It relates particularly to magnetic resonance imaging apparatus employing "short" bore magnets and will be described with particular reference thereto. However, the present invention finds application in any magnetic imaging apparatus wherein a gradient coil is disposed in a magnetic field such that Lorentzian force components on the coil are not balanced.

Magnetic resonance imagers commonly include a bore having a diameter of 90 centimetres or more for receiving the body of an examined patient. The bore is surrounded by a series of annular superconducting magnets for creating a substantially uniform magnetic field longitudinally along the patient receiving bore. The more axially spaced the annular magnets, the more uniform the primary magnetic field within the patient receiving bore tends to be and the longer the axial dimension over which such magnetic field uniformity exists. Typically, the bore is at least 1.6 metres long and often longer.

One of the drawbacks to such "long" bore magnets is that the region of interest is often inaccessible to medical personnel. If a procedure is to be performed based on the image, the patient must be removed from the bore before the procedure can be performed. Moving the patient risks potential misregistration problems between the image and the patient.

One way to improve access to the patient is to shorten the length of the magnet and the patient receiving bore. If the magnet and the bore were shortened to about 1 metre or roughly the diameter of the bore, patient access is much improved. Although the size of the uniform magnetic field area compresses to a more disk-like shape, the area of substantial uniformity is still sufficient for a series of 10 to 20 contiguous slice images. NMR helical or continuous scanning methods can also be employed.

Although an adequate imaging volume remains, the magnetic field in the volume around the periphery of the bore which receives the gradient coil tends to become relatively non-uniform and has both axial z-components and radial x,y-components. The gradient coil generally includes windings for generating three linear and orthogonal magnetic field gradients for providing spatial resolution and discrimination of nuclear magnetic resonance signals. Gradient coils are typically designed and constructed to optimize strength and linearity over the imaging volume and stored energy and inductance in the gradient coil. See, for example, U.S. Patent No. 5,296,810 of Morich. To create the magnetic field gradients, current pulses are applied to the x, y, and/or z-gradient coils. These currents interact with the main magnetic field to generate Lorentz forces on the gradient coil. Due to the symmetries included in gradient coil currents, the Lorentz forces across the entire coil cancel when the gradient coil is disposed in a uniform magnetic field. However, when the main magnetic field is less uniform, particularly when there are significant radial and non-uniform axial components in the neighborhood of the gradient coils, a net thrust can be developed. Typically, pulsing the z-gradient coil causes a net thrust in the z-direction, pulsing the x-gradient coil develops a net thrust in the x-direction, and pulsing the y-gradient coil causes a net thrust in the y-direction. In the case of the z-gradient coil, the net axial force can be on the order of a few hundred pounds. These net thrusts tend to push or urge the gradient coil axially out of the bore. Although the gradient coil can be anchored mechanically, these large forces still tend to cause acoustic noise and increased vibrations to the magnet and the patient. Such vibration has deleterious effects on imaging, such as a loss of resolution.

The present invention provides a magnetic resonance imaging apparatus including a gradient coil which alleviates the above-referenced problems.

According to the invention there is provided a magnetic resonance imaging apparatus which includes a main magnet for generating a temporally constant magnetic field through a patient receiving bore and a gradient field coil assembly and an RF coil around the bore and in which the temporally constant magnetic field has components which interact with current pulses applied to principal windings of the gradient field coil assembly to generate-magnetic field gradients in such a manner that a net force is experienced by the gradient field coil assembly, characterised by force correction windings mounted adjacent edge areas of the gradient field coil assembly which carry currents in an opposite sense to the principal windings such that the net force is substantially cancelled.

In an apparatus according to the invention the patient receiving bore typically has a length to a diameter ratio of less than 1.5:1, e.g. of substantially 1:1.

In one particular embodiment of the invention the gradient coil assembly is a self-shielded gradient coil having a primary coil and a shield coil, current pulses passing through the primary and shield coils generating magnetic fields that combine within the gradient coil assembly to form a linear magnetic field gradient and tending to cancel outside the gradient coil assembly, and the current pulses that pass through the shield coil interact with radial magnetic field components to generate a first force, the primary gradient coil being wound such that the current pulses passing through the primary gradient coil interact with the radial magnetic field components to generate a second force that is substantially equal and opposite to the first force.

In another particular embodiment the gradient coil assembly includes primary and secondary gradient coils which each include four symmetrically arranged thumbprint coils for generating a magnetic field gradient transverse to a longitudinal axis of the bore, the thumbprint coils further generating a net transverse force, the force correction windings

including force offsetting current loops disposed adjacent an end of at least one of the primary and secondary gradient coils for carrying current flows which circulate in an opposite direction to a most adjacent thumbprint coil for offsetting the net transverse force along the transverse direction.

In a further particular embodiment the gradient coil assembly includes a primary z-gradient coil including a series of loop coils which are mounted around a cylindrical former for generating a z-gradient field and the force correction windings include loops extending around ends of the cylindrical former and connected in series with the z-gradient loop coils for carrying common current pulses but in an opposite direction.

In a still further embodiment the gradient coil assembly includes a primary z-gradient coil and a secondary, shield z-gradient coil and the secondary z-gradient coil is disposed surrounding the primary z-gradient coil, current pulses applied to the primary and secondary z-gradient coils causing magnetic fields which combine within the gradient coil assembly to generate a linear gradient axially along the coil assembly and which generally cancel exterior to the coil assembly.

The present invention also provides a method of designing gradient coils for a magnetic resonance imaging apparatus in which coil windings patterns are constrained to have a preselected gradient field strength, at least a preselected gradient field linearity and at least a preselected gradient field uniformity, characterised by: constraining the coil windings such that the gradient coil assembly experiences substantially no axial or transverse force when current pulses flowing through the gradient coil windings interact with radial and axial components of a main magnetic field within which the gradient coil is disposed.

One advantage of the present invention is that it facilitates access to portions of the patient in the examination region.

Another advantage of the present invention is that it improves image quality.

Another advantage of the present invention is that it simplifies mounting and construction of gradient coils.

One magnetic resonance apparatus in accordance with the invention and a method in accordance with the invention of designing the coils of a gradient coil assembly of the apparatus will now be described, by way of example, with reference to the accompanying drawings in which:

FIGURE 1 is a diagrammatic illustration of the magnetic resonance imaging apparatus.

FIGURES 2A and 2B illustrate exemplary current distributions for z-primary and shield gradient coils of the apparatus; and,

FIGURES 3A and 3B illustrate a primary and shield or secondary x or y-gradient coil winding of the apparatus.

Referring to FIGURE 1, the apparatus includes a plurality of primary magnet coils 10 which generate a temporally constant magnetic field along a longitudinal or z-axis of a central bore 12. The primary magnet coils are supported by a former 14 and received in a toroidal helium vessel or can 16. The can 16 is filled with liquid helium to maintain the primary magnet coils at superconducting temperatures. The can 16 is surrounded by one or more cold shields 18 which are supported in a vacuum dewar 20.

A whole body gradient coil assembly 30 includes x, y, and z-coils mounted around the bore 12. Preferably, the gradient coil assembly is a self-shielded gradient coil assembly that includes primary x, y, and z-coil assemblies 32 potted in a dielectric former 34 and a secondary or shield gradient coil assembly 36 that is supported on a bore defining cylinder 38 of the vacuum dewar 20. The dielectric former 34 with potted gradient coils can function as a bore liner or a cosmetic bore liner can be inserted to line it. Preferably, shims (not shown) for adjusting the magnetic field are positioned as needed between the primary and shield coil dielectric formers as taught by DeMeester, et al. in U.S. Patent No. 5,349,297. A whole body RF coil 40 is mounted inside the gradient coil assembly 30. A whole body RF shield 42, e.g. a layer of copper mesh, is mounted between RF coil 40 and the gradient coil assembly 30.

An operator interface and control station 50 includes a human-readable display such as a video monitor 52 and an operator input means including a keyboard 54 and a mouse 56. Track balls, light pens, and other operator input devices are also contemplated. Computer racks 58 hold a magnetic resonance sequence memory and controller, a reconstruction processor, and other computer hardware and software for controlling the radio frequency coil 40 and the gradient coil 30 to implement any of a multiplicity of conventional magnetic resonance imaging sequences, including echo-planar, echo-volume, spin echo, and other imaging sequences. Echo-planar and echo-volume imaging sequences are characterized by short data acquisition times and high gradient strengths and slew rates. The racks 58 also hold a digital transmitter for providing RF excitation and resonance manipulation signals to the RF coil and a digital receiver for receiving and demodulating magnetic resonance signals. An array processor and associated software reconstruct the received magnetic resonance signals into an image representation which is stored in computer memory, on disk, or in other recording media. A video processor selectively extracts portions of the stored reconstructed image representation and formats the data for display by the video monitor 52. An image printer can also be provided for making paper copies of selected images.

In the preferred embodiment, the diameter and length of the bore 12 have a size ratio of about 1:1. However, it is to be appreciated that the present invention is also applicable to other magnets, particularly magnets with limited main magnetic field uniformity. Typically, one might expect magnets with a bore length to diameter ratio of up to 1.5:1 to be candidates for the present invention. However, the present invention will also be applicable to magnetic resonance imagers with longer bores in which there is a sufficient main magnetic field non-uniformity, particularly radial field components, in the area of the gradient coils to cause image degradation due to vibration or in which there are sufficient net forces that mechanical mounting of the gradient coils becomes difficult.

The main magnetic field magnets 10 produce a main magnetic field B_0 over the imaging volume. In the short bore magnets, the z-component of the main magnetic field B_{z0} remains constant over a central imaging region of about 40 to 45 centimetres in diameter in a x,y plane at the magnetic isocenter. Along the z-axis, the imaging region is shorter than 40 centimetres. Outside the ellipsoidal, uniform, imaging region, the B field changes significantly at larger axial or z-displacements from isocenter and with radial or ρ displacement from the central axis of the bore. The dependence of B_{z0} on the axial and radial positions becomes more significant around the gradient and RF coil locations, typically the region bounded by 30 cm. $\leqq \rho \leqq$ 40 cm. and -60 cm. $\leqq z \leqq$ 60 cm and encompassing all angular positions. Although inside the imaging region, the contribution of the radial component of the main magnetic field $B_{\rho 0}$ to the total value of the B_0 field is negligible, it is much more significant at the gradient coil sets' location, particularly in short bore magnet designs.

The interaction between the spatially varying components of the magnetic fields and the current densities of the coil set creates two distinct problems. First, there is an interaction between the azimuthally directed current of the z-gradient coil with $B_{\rho 0}$. Because both quantities are odd symmetric functions around the geometric center of the magnet plus gradient system, they generate a z-directed thrust force according to the Lorentz force equation. Depending on the axial or z-behavior of $B_{\rho 0}$, in particular its value at the location of the conductors of the z-gradient coil, the value of the thrust force can reach several hundred pounds or a few thousand Newtons.

The second problem deals with the interaction between the current density for a transverse x or y-gradient coil and both the $B_{\rho 0}$ and B_{z0} components of the main magnetic field. The result of such an interaction is again determined from the Lorentz force equation, but is a radially directed net force. Due to the azimuthal dependence of the transverse gradient's current density, the radially directed force is along the x or y-direction for an x or y-gradient coil, respectively. Again, the magnitude of this force is dependent on the value of the static magnetic fields $B_{\rho 0}$ and B_{z0} components at the locations of the current lines for the x and y-gradient coils. Because the radius of the x and y-gradient coils is different, the value of the corresponding force will also be different. The value of the components of B0 change with radial position.

Looking first to the analytical evaluation of the z-directed thrust force which is the result of the interaction between the z-gradient coil and the $B_{\rho 0}$ component, a minimization technique is presented for designing z-gradient coils with a zero net thrust force. For a conventional z-gradient coil, the current is odd-symmetric around the geometric center of the coil and the magnetic field $B_{\rho 0}$ is also odd-symmetric around the geometric center of the magnet. Because these two centers normally coincide, the result of the interaction between the current and the $B_{\rho 0}$ field component is a z-directed thrust force on the gradient coil.

With reference to FIGURES 2A and 2B, the z-gradient coil is self-shielded, having an inner or primary winding 60 of radius a and an outer or secondary shield coil 62 of radius b. For simplicity of construction, the primary z-gradient coil is preferably circular loops wound on the dielectric former 34 and potted in epoxy or other resin. The shield z-gradient coil is preferably circularly wound on the dielectric former 38 that is incorporated into the vacuum dewar. Of course, alternative embodiments where it is not an integral part of the vacuum dewar are also possible. The current density for the inner gradient coil $J_\phi^a(z)$ is azimuthally directed and varies only along the axial direction of the coil. Similarly, the current density for the outer gradient coil $J_\phi^b(z)$ is azimuthally directed and also varies along the axial direction of the coil. The interaction between these two current densities with the radial component of the main magnetic field at the radial locations of the coil, i.e., $B_{\rho 0}^a$ and $B_{\rho 0}^b$, respectively, generates a Lorentz force which is directed along the z-direction.

In general, the Lorentz equation describes the elemental force experienced by a current element $J_\phi^{a,b}(z)dz(a,b)$ $d\phi\hat{a}_\phi$ in the presence of the magnetic field $B_{\rho 0}^{a,b}$ as:

$$d\vec{F} = j_\phi^a(z)\,dz\,a\,d\phi\alpha_\phi \times B_{\rho 0}^a\alpha_\rho + j_\phi^b(z)\,dz\,b\,d\phi\alpha_\phi \times B_{\rho 0}^b\alpha_\rho \qquad (1).$$

The net force which is generated from these The net force which is generated from these two coils is along the z-direction and has the form:

$$\vec{F}^{net} = F_z^{net}\alpha_z$$

with

$$F_z^{net} = -2\pi\left\{a\int_{-\infty}^{+\infty} j_\phi^a(z)\, B_{\rho 0}^a(z)\, dz + b\int_{-\infty}^{+\infty} j_\phi^b(z)\, B_{\rho 0}^b(z)\, dz\right\} \qquad (2),$$

where $B_{\rho 0}{}^{a,b}(z)$ are known real functions of (z). The Fourier transform pairs for $j_\phi{}^{a,b}$ and $B_{\rho 0}{}^{a,b}$ are:

$$j_\phi^{a,b}(k) = \int_{-\infty}^{+\infty} j_\phi^{a,b}(z)\, e^{-ikz} dz \qquad (3a),$$

$$j_\phi^{a,b}(z) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} j_\phi^{a,b}(k)\, e^{+ikz} dk \qquad (3b),$$

$$B_{\rho 0}^{a,b}(k) = \int_{-\infty}^{+\infty} B_{\rho 0}^{a,b}(z)\, e^{-ikz} dz \qquad (3c),$$

$$B_{\rho 0}^{a,b}(z) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} B_{\rho 0}^{a,b}(k)\, e^{+ikz} dk \qquad (3d).$$

These equations can be used to obtain the Fourier or k-space representation of the net thrust force as:

$$F_z^{net} = \left\{a\int_{-\infty}^{+\infty} j_\phi^a(k)\, B_{\rho 0}^a(k)\, dk + b\int_{-\infty}^{+\infty} j_\phi^b(z)\, B_{\rho 0}^b(k)\, dk\right\} \qquad (4).$$

Because $B_{\rho 0}{}^{a,b}(z)$ is a pure real, odd-symmetric function of z:

$$B_{\rho 0}^{*(a,b)}(k) = \int_{-\infty}^{+\infty} B_{\rho 0}^{a,b}(z)\, e^{+ikz} dz = -B_{\rho 0}^{a,b}(k) \qquad (5).$$

For a set of z-gradient coils with radius a,b, respectively, the expression of the axial z-component of the magnetic field due to the gradient coil B_z, in the three regions which the two coils define can be simplified, due to the behavior of the current density. More specifically, no angular dependence is considered for a current which is uniform around the circumference of a cylindrical gradient coil. The current in the z-gradient coil is restricted to vary along the z-direction. With these constraints, the expression of B_z in the three regions is:

$$B_z = \frac{\mu_0}{2\pi}\int_{-\infty}^{\infty} dk\, e^{ikz} k\left[a j_\phi^a(k) K_1(ka) I_0(k\rho) + b j_\phi^b(k) K_1(kb) I_0(k\rho)\right] \quad \text{for } \rho<a \qquad (6a),$$

$$B_z = \frac{\mu_0}{2\pi}\int_{-\infty}^{\infty} dk\, e^{ikz} k\left[-a j_\phi^a(k) K_0(k\rho) I_1(ka) + b j_\phi^b(k) K_1(kb) I_0(k\rho)\right] \quad \text{for } a<\rho<b \qquad (6b),$$

$$B_z = \frac{\mu_0}{2\pi}\int_{-\infty}^{\infty} dk\, e^{ikz} k\left[-a j_\phi^a(k) K_0(k\rho) I_1(ka) - b j_\phi^b(k) K_0(k\rho) I_1(kb)\right] \quad \text{for } b<\rho \qquad (6c),$$

where $j_\phi{}^a(k)$, $j_\phi{}^b(k)$ are the Fourier transforms of the current densities of the inner coil of radius a and the outer coil of radius b, respectively. Ideally, the z-gradient coil generated magnetic field B_z is linear in the internal region of both coils and zero outside the two coils. In order to satisfy the constraint that the field is zero outside of the two coils, B_z in the region b<ρ is set to zero, i.e., Equation (6c) is set to zero. One way to set Equation (6c) to zero is by relating the current densities of the inner and outer coils such that their Fourier components sum to zero, i.e.:

$$a j_\phi^a(k)\, K_0(k\rho)\, I_1(ka) + b j_\phi^b(k)\, K_0(k\rho)\, I_1(kb) = 0 \quad (7).$$

or:

$$j_\phi^b(k) = -\frac{a I_1(ka)}{b I_1(kb)}\, j_\phi^a(k) \quad (8).$$

Substituting Equation (8) into Equation (6a), the expression for the magnetic field for the region inside of the two coils $\rho<a$ becomes:

$$B_z = \frac{\mu_0 a}{2\Pi}\int_{-\infty}^{\infty} dk\, k\, e^{ikz} j_\phi^a(k)\, I_0(k\rho)\, K_1(ka)\left(1 - \frac{I_1(ka)\, K_1(kb)}{I_1(kb)\, K_1(ka)}\right) \quad (9).$$

From Equation (8), the net force expression can also be simplified:

$$F_z^{net} = a\left\{\int_{-\infty}^{+\infty} j_\phi^a(k)\left(B_{\rho 0}^a(k) - \frac{I_1(ka)}{I_1(kb)} B_{\rho 0}^b(k)\right)\right\} dk \quad (10).$$

The stored energy in the coil W in terms of Equation (8) is:

$$W = \frac{a^2 \mu_0}{2}\int_{-\infty}^{\infty} dk\, |j_\phi^a(k)|^2\, I_1(ka)\, K_1(ka)\left(1 - \frac{I_1(ka)\, K_1(kb)}{I_1(kb)\, K_1(ka)}\right) \quad (11).$$

For a gradient coil of length L, the Fourier expansion around the geometric center for the current of the inner coil in a self-shielded design is:

$$j_\phi^a(z) = \sum_{n=1}^{\infty} j_n^a \sin k_n z \quad \text{for } |z| \le \frac{L}{2} \quad (12),$$

$$j_\phi^a(z) = 0 \text{ for } |z| > \frac{L}{2} \quad (13),$$

where j_n^a are the Fourier coefficients of the expansion and $\sin k_n z$ represents the antisymmetry condition of the current around the origin. For a coil of length L, the current is preferably restricted to become zero at the ends of the coil which restricts the values that k_n can take. Thus, the allowable values for k_n are:

$$j_\phi^a(z) = 0 \text{ for } |z| = \frac{L}{2} \Rightarrow k_n = \frac{2n\Pi}{L} \quad (14).$$

The Fourier transform of $j_\phi^a(z)$ is:

$$j_\phi^a(m,k) = \sum_{n=1}^{\infty} i\, \frac{L}{2}\, j_n^a\, \psi_n(k) \quad (15),$$

with

$$\psi_n(k) = \left[\frac{-\sin(k-k_n)\frac{L}{2}}{(k-k_n)\frac{L}{2}} + \frac{\sin(k+k_n)\frac{L}{2}}{(k+k_n)\frac{L}{2}}\right] \quad (16).$$

Due to the symmetry requirements, the dependence of $\Psi_n(k)$ in the variable k is:

$$\Psi_n(-k) = -\Psi_n(k) \quad (17).$$

From the expression of the Fourier component of the current, the expressions for the gradient magnetic field B_z, the stored magnetic energy W in the gradient coil, and the z-directed thrust force F_z are:

$$B_z = -\frac{\mu_0 a}{2\pi}\int_{-\infty}^{\infty} dk \sin kz\, k\sum_{n=1}^{\infty} \frac{L}{2} j_n^a \psi_n(k)$$
$$I_0(k\rho)K_1(ka)\left(1-\frac{I_1(ka)K_1(kb)}{I_1(kb)K_1(ka)}\right) \qquad (18),$$

$$W_m = \frac{a^2\mu_0}{2}\int_{-\infty}^{\infty} dk I_1(ka)K_1(ka)$$
$$\left(1-\frac{I_1(ka)K_1(kb)}{I_1(kb)K_1(ka)}\right)\sum_{n=1}^{\infty}\sum_{n'=1}^{\infty} j_n^a j_{n'}^a \frac{L^2}{4}\psi_n(k)\psi_{n'}(k) \qquad (19).$$

$$F_z^{net} = a\sum_{n=1}^{\infty} i\frac{L}{2} j_n^a \int_{-\infty}^{\infty} dk\psi_n(k)\left\{B_{\rho 0}^a(k) - \frac{I_1(ka)}{I_1(kb)}B_{\rho 0}^b(k)\right\} \qquad (20).$$

Because $B_{\rho 0}{}^{a,b}(z)$ is an odd-symmetric function with respect to z, its Fourier transform is defined as:

$$B_{\rho 0}^{a,b}(k) = -i\int_{-\infty}^{\infty} dz(\sin kz)B_{\rho 0}^{a,b}(z)$$
$$= -i\tilde{B}_{\rho 0}^{a,b}(k) \qquad (21a),$$

with:

$$\tilde{B}_{\rho 0}^{a,b}(k) = \int_{-\infty}^{\infty} dz(\sin kz)B_{\rho 0}^{a,b}(z) \qquad (21b).$$

Equation (20) then has the form:

$$F_z^{net} = a\sum_{n=1}^{\infty} \frac{L}{2} j_n^a \int_{-\infty}^{\infty} dk\psi_n(k)\left\{\tilde{B}_{\rho 0}^a(k) - \frac{I_1(ka)}{I_1(kb)}\tilde{B}_{\rho 0}^b(k)\right\} \qquad (22).$$

From the expressions for the magnetic field, the stored energy, and the z-directed thrust, the functional E can be defined as:

$$E(j_n^a) = W - \sum_{j=1}^{N_1}\lambda_j(B_z(\vec{r}_j) - B_{zSC}(\vec{r}_j)) - \sum_{j=N_1+1}^{N_1+1}\lambda_j\left(F_z^{net} - F_{zSC}^{net}\right) \quad (23),$$

where B_{zSC} and $F_{zSC}{}^{net}$ are the prespecified constraint values of the magnetic field at the constraint points and the z-directed thrust force, respectively.

Minimizing E with respect to $j_n{}^a$, one obtains a matrix equation for the $j_{n'}{}^a$ as follows:

$$\sum_{n'=1}^{\infty} j_{n'}^a\left\{\frac{aL\pi}{2}\int_{-\infty}^{\infty} dk I_1(ka)K_1(ka)\left(1-\frac{I_1(ka)K_1(kb)}{I_1(kb)K_1(ka)}\right)\psi_n(k)\psi_{n'}(k)\right\}$$
$$= -\sum_{j=1}^{N_1}\lambda_j\int_{-\infty}^{\infty} dk\, k\sin kz_j I_0(k\rho_j)K_1(ka)\left(1-\frac{I_1(ka)K_1(kb)}{I_1(kb)K_1(ka)}\right)\psi_n(k)$$
$$-\frac{2\pi}{\mu_0}\sum_{j=N_1+1}^{N_1+1}\lambda_j\int_{-\infty}^{\infty} dk\psi_n(k)\left\{\tilde{B}_{\rho 0}^a(k) - \frac{I_1(ka)}{I_1(kb)}\tilde{B}_{\rho 0}^b(k)\right\} \qquad (24a),$$

or

$$\sum_{n'=1}^{\infty} j_{n'}^{a} C_{n'n} = \sum_{j=1}^{N_1+1} \lambda_j D_{jn} \qquad (24b).$$

Depending on the value of j, the D_{jn} matrix can correspond to either the magnetic field expression or to the z-directed thrust force expression.

Specifically, the conversion for the D_{jn} matrix is:

$$D_{jn} = \begin{cases} \text{B-field for } j = 1, N_1 \\ F_z^{net} \quad \text{for } j = N_1 + 1, N_1 + 1 \end{cases} \qquad (25).$$

For the expression of $B_{\rho 0}{}^{a,b}(z)$, data is obtained from the actual main magnet 10. Due to the antisymmetry of the $B_{\rho 0}{}^{a,b}(z)$, only positive z-values are generated. In order to incorporate the generated data for the radial component of the magnetic field into Equation (24a), consider the following:

$$B_{\rho 0}^{a,b}(z) = \sum_{\gamma=0}^{\Omega} B_{\gamma}^{a,b}(z_\gamma)\,\delta(z - z_\gamma) \qquad (26),$$

where Ω corresponds to the total number of points along z, and $B_\gamma{}^{a,b}$ is the corresponding value of $B_{\rho 0}{}^{a,b}(z)$ at the location z_γ. The Fourier transform $B_{\rho 0}{}^{a,b}(k)$, is:

$$\tilde{B}_{\rho 0}^{a,b}(k) = 2\sum_{\gamma=0}^{\Omega} B_{\gamma}^{a,b}(z_\gamma) \sin k z_\gamma \Delta z_\gamma \qquad (27).$$

Furthermore, the generated values for $B_{\rho 0}{}^{a,b}(z)$ are for the axial distances which are larger than the half length of the gradient coil and up to the point where there is no significant action between the current density of the gradient coils and the main magnet's radial components. Truncating the infinite summations at M terms, the matrix representation for Equation (24b) becomes:

$$\underline{J}^a \underline{C} = \underline{\lambda}\ \underline{D} \text{ or } \underline{J}^a = \underline{\lambda}\ \underline{D}\ \underline{C}^{-1} \qquad (28)$$

where $\underline{J}^a$ is a 1xM matrix, $\underline{C}$ is a MxM matrix, $\underline{\lambda}$ is a 1xN_1+1 matrix, and $\underline{D}$ is an N_1+1xM matrix.

Using the constraint equation for the gradient magnetic field B_z and the z-directed thrust force, the Lagrangian multipliers are determined. The Fourier components of the current density for the primary or inner gradient coil in matrix form are:

$$\underline{J}^a = \underline{B}_z[\underline{D}\ \underline{C}^{-1}\underline{D}^t]^{-1}\underline{D}\ \underline{C}^{-1} \qquad (29).$$

Finding the expression for the Fourier components of the current for the inner coil and with the help of Equation (14), the continuous distribution of current density of the inner coil is generated. To determine the current for the outer coil, the shielding relationship between the Fourier transform for both current densities of Equation (8) is used. An inverse Fourier transform is then used to obtain the continuous current distribution for the outer coil.

The next step is the process for discretization of the continuous current distribution for both coils. The continuous current distribution is divided into positive and negative current regions. Integrating the area underneath each region, the total current contained in each region is obtained. When the current for all of the regions of the cylinder is calculated, discrete current loops are positioned on a dielectric former to mimic the behavior of the continuous current pattern. Each region is filled with discrete wires carrying the prescribed amount of current. The amount of current is the same for each wire loop, in the preferred embodiment. In each region, the continuous current density is divided into smaller segments which correspond to the selected equal amount of current. The selected current amount may be iteratively adjusted in order to match the continuous current densities in the selected current regions more closely. Each wire is placed at the midpoint of the corresponding segment in order to obtain an equal distribution from both sides of the segment. This current distribution is then analyzed to calculate the generated magnetic field to assure that the desired magnetic field is, in fact, obtained. Alternately, one can use the center of mass scheme taught in previously referenced U.S. Patent No. 5,296,810.

With continuing reference to FIGURE 2, in the preferred embodiment, the inner or primary coil has a radius of about 340 millimetres and a length of about 700 to 900 millimetres. The outside or shield coil is about 385 millimetres in diameter. For the Fourier series expansion, ten points provide a reasonable definition of the current density, although larger or smaller numbers may be chosen. Five constraint points are preferably chosen to define the characteristics of

the field and the thrust force. The field is to be constrained inside a 40 cm. to 50 cm. generally ellipsoidal imaging volume. The first constraint point establishes a gradient field strength, e.g., about 13.5 mT/m. The second constraint point defines the linearity of the field along the gradient axis. Preferably, at the edges of the 25 cm. dimension of the volume, the magnetic field is confined to vary not more than 5%. The remaining two constraints define the uniformity of the field across a plane perpendicular to the gradient axis. Preferably, the magnetic field is constrained to stay within 7% of its actual value at a radial distance of about 22 cm. from the center of the coils.

The last constraint point defines the value of the z-directed thrust force which is preferably less than $-1.0e^{-08}$ Newtons. A suitable primary gradient coil meeting these constraints is illustrated in FIGURE 2A and a suitable outer or shield gradient coil meeting these conditions is illustrated in FIGURE 2B. It will be noted that the primary gradient coil has several force correcting windings 64 of reversed polarity at its ends or extremes. If these reverse polarity windings were removed, the gradient coil would suffer a net longitudinal force component in excess of the above-discussed constraints.

Although the above-described method calculates an ideal current distribution, the force correcting current for cancelling the axial force can be determined iteratively. More specifically, several coil windings are disposed at the ends of one of the primary and secondary coils, preferably the primary coils. The net force on the gradient coil assembly during a gradient current pulse is measured. The current flow through the force correcting windings (or the number of force correcting loops) is iteratively adjusted until the axial force is substantially zeroed.

With reference to FIGURES 3A and 3B, the gradient coil assembly further includes an x-gradient coil and a y-gradient coil. The x and y-gradient coils are of substantially the same construction, but rotated 90°. Of course, because one is laminated over the other, they will have a small difference in radius. The x and y-gradient coils each include four substantially identical windings arranged symmetrically on either side of the isocenter. Each of the quadrants of the primary x or y-gradient coil contains a winding substantially as illustrated in FIGURE 3A. Each of the quadrants of the shield x or y-gradient coil contains a winding substantially as illustrated in FIGURE 3B. A current distribution is again calculated to produce the selected gradient strength, minimize stored energy, and zero the net lateral Lorentz forces.

For the transverse x or y-gradient coils, the total current density can be represented as:

$$\vec{J}^{\,a}(\vec{r}) = \left[j_\phi^a(\phi,z)\,\mathfrak{a}_\phi + j_z^a(\phi,z)\,\mathfrak{a}_z\right]\delta(\rho-\rho_0) \qquad (30),$$

where $\delta(\rho-\rho_0)$ is the restriction that the current is confined on a cylindrical surface of radius ρ_0. Again, the x and y-gradient coils are self-shielded. That is, they have an inner or primary coil of radius a and an outer secondary or shield coil of radius b. The current density for the outer coil of radius b is analogous to Equation (30). The interaction between the components of the current density and the corresponding components of the main magnetic field yields an x-directed Lorentz force for self-shielded x-gradient coils and a y-direction force for self-shielded y-gradient coils. The following discussion focuses on x-gradient coils. However, it is to be appreciated that the same discussion is equally applicable to y-gradient coils which will be of the same construction but rotated 90° and of slightly different radius.

The Lorentz equation describes the elemental force experienced by a current element Id1 in the presence of the magnetic field B_0:

$$d\vec{F} = I\vec{d}1 \times \vec{B}_0 \qquad (31a),$$

or

$$d\vec{F} = \left(j_\phi^{a,b}(z)\cos(\phi)\,\mathfrak{a}_\phi + j_z^{a,b}(z)\sin(\phi)\,\mathfrak{a}_z\right)dz\,(a,b)\,d\phi \times \left(B_{\rho 0}^{a,b}(z)\,\mathfrak{a}_r + B_{z0}^{a,b}(z)\,\mathfrak{a}_z\right) \qquad (31b),$$

where $B_{\rho 0}{}^{a,b}(z)$ and $B_{z0}{}^{a,b}(z)$ are radial and axial components of the main magnetic field B_0 at radii locations (a,b). Integrating Equation (31b) results in a net force along the x-axis:

$$\vec{F}^{net} = F_x^{net}\,\mathfrak{a}_x \qquad (32a),$$

with:

$$F_x^{net} = 4\pi a\int_0^{+\infty}\{j_\phi^a(z)B_{z0}^a(z) - j_z^a(z)B_{\rho 0}^a(z)\}dz + 4\pi b\int_0^{+\infty}\{j_\phi^b(z)B_{z0}^b(z) - j_z^b(z)B_{\rho 0}^b(z)\}dz \qquad (32b),$$

where $B_{\rho 0}{}^{a,b}(z)$ and $B_{z0}{}^{a,b}(z)$ are known real functions of z. The Fourier transform pairs for $j_\phi{}^{a,b}$, $j_z{}^{a,b}$, $B_{\rho 0}{}^{a,b}$, and $B_{z0}{}^{a,b}$ are:

$$j_\phi^{a,b}(k) = \int_{-\infty}^{+\infty} j_\phi^{a,b}(z)\, e^{-ikz} dz \qquad (33a),$$

$$j_\phi^{a,b}(z) = \frac{1}{2\Pi}\int_{-\infty}^{+\infty} j_\phi^{a,b}(k)\, e^{+ikz} dk \qquad (33b),$$

$$j_z^{a,b}(k) = \int_{-\infty}^{+\infty} j_z^{a,b}(z)\, e^{-ikz} dz \qquad (33c),$$

$$j_z^{a,b}(z) = \frac{1}{2\Pi}\int_{-\infty}^{+\infty} j_z^{a,b}(k)\, e^{+ikz} dk \qquad (33d),$$

$$B_{\rho 0}^{a,b}(k) = \int_{-\infty}^{+\infty} B_{\rho 0}^{a,b}(z)\, e^{-ikz} dz \qquad (33e),$$

$$B_{\rho 0}^{a,b}(z) = \frac{1}{2\Pi}\int_{-\infty}^{+\infty} B_{\rho 0}^{a,b}(k)\, e^{+ikz} dk \qquad (33f),$$

$$B_{z0}^{a,b}(k) = \int_{-\infty}^{+\infty} B_{z0}^{a,b}(z)\, e^{-ikz} dz \qquad (33g),$$

$$B_{z0}^{a,b}(z) = \frac{1}{2\Pi}\int_{-\infty}^{+\infty} B_{z0}^{a,b}(k)\, e^{+ikz} dk \qquad (33h),$$

Using the constraint:

$$\vec{\nabla} \cdot \vec{J} = 0$$

and the shielding conditions, the following relationships are derived:

$$j_z^{a,b}(k) = -\frac{1}{k(a,b)} j_\phi^{a,b}(k) \qquad (34a),$$

$$j_\phi^{b}(k) = -\frac{a}{b}\,\frac{I'_1(ka)}{I'_1(kb)}\, j_\phi^{a}(k) \qquad (34b),$$

The expression for F_x^{net} from Equation (32b) is rewritten as:

$$F_x^{net} = 2a\int_0^{+\infty} j_\phi^{a,b}(k)\left\{\left[B_{z0}^{*a}(k) + \frac{1}{ka}B_{\rho 0}^{*a}(k)\right] - \frac{I'_1(ka)}{I'_1(kb)}\left[B_{z0}^{*b}(k) + \frac{1}{kb}B_{\rho 0}^{*b}(k)\right]\right\} \qquad (35),$$

where * represents the complex conjugate of the Fourier transforms of the two components of the main magnetic field.

For a real magnet design, there is no useful analytic expression for the $B_{z0}^{a,b}(z)$ and $B_{\rho 0}^{a,b}(z)$. In order to incorporate analytically the numerically generated data for both components of the magnetic field along the central z-axis of the magnet, consider the following representations:

$$B_{z0}^{a,b}(k) = 2\sum_{\gamma=0}^{\Omega} B_{z\gamma}^{a,b}(z_\gamma)\cos kz_\gamma \Delta z_\gamma = B_{z0}^{*(a,b)}(k) \qquad (36a),$$

$$B_{\rho 0}^{a,b}(k) = 2\sum_{\gamma=0}^{\Omega} B_{\rho\gamma}^{a,b}(z_\gamma)\sin kz_\gamma \Delta z_\gamma - B_{\rho 0}^{*(a,b)}(k) \qquad (36b),$$

where $B_{z0}^{a,b}(z)$ is symmetric around z and $B_{\rho 0}^{a,b}(z)$ is antisymmetric around z.

The restriction to the inner coil length, the confinement of the current density to a cylindrical surface, the azimuthal and axial symmetries for the j_ϕ^a and j_z^a and the demand that the current density obeys the continuity equation, provides a Fourier series expansion for both components around the geometric center of the coil:

$$j_\phi^a(\phi, z) = \cos\phi \sum_{n=1}^{\infty} j_{\phi n}^a \cos k_n z \qquad \text{for } |z| \le \frac{L}{2} \qquad (37a),$$

$$j_z^a(\phi, z) = \sin\phi \sum_{n=1}^{\infty} \frac{j_{\phi n}^a}{k_n a} \sin k_n z \qquad \text{for } |z| \le \frac{L}{2} \qquad (37b),$$

where $j_{\phi n}^a$ are the Fourier coefficients, L represents the total length of the inner coil, and $k_n = 2n\Pi/L$ because the current cannot flow off of the ends of the cylinder. Furthermore, both current components are zero for $|z| > L/2$.

$$B_z = -\frac{\mu_0 a}{2\Pi} \sum_{m=-\infty}^{\infty} e^{+im\phi} \int_{-\infty}^{\infty} dk\, k\, e^{ikz} j_\phi^a(m,k)\; I_m(k\rho) K'_m(ka) \left(1 - \frac{I'_m(ka) K'_m(kb)}{I'_m(kb) K'_m(ka)}\right) \qquad (38),$$

The general expression for the magnetic field for a self-shielded gradient coil in terms of the Fourier transform of the current density is: where $j_\phi^a(m, k)$ is the double Fourier transform of $J_\phi^a(\phi, z)$. Since the azimuthal dependence of j_ϕ^a is proportional to $\cos(\phi)$, the Fourier transform of j_ϕ^a is non-zero when $m = \pm 1$. In this case the two-dimensional Fourier transform of the current density is:

$$j_\phi^a(\pm 1, k) = \frac{1}{2} \sum_{n=1}^{\infty} \frac{L}{2} j_{\phi n}^a \psi_n(k) \qquad (39),$$

with:

$$\psi_n(k) = \left[\frac{\sin(k-k_n)\frac{L}{2}}{(k-k_n)\frac{L}{2}} + \frac{\sin(k+k_n)\frac{L}{2}}{(k+k_n)\frac{L}{2}} \right] \qquad (40),$$

where $\Psi_n(k)$ is an even function of k and $j_\phi^a(+1,k) = j_\phi^a(-1,k)$. Therefore, the expression of the gradient field has the form:

$$B_z = -\frac{\mu_0 a L}{4\Pi} \cos\phi \sum_{n=1}^{\infty} j_{\phi n}^a \int_{-\infty}^{\infty} dk \cos kz\, \psi_n(k)\, k\; I_1(k\rho) K'_1(ka) \left(1 - \frac{I'_1(ka) K'_1(kb)}{I'_1(kb) K'_1(ka)}\right) \qquad (41).$$

In a similar fashion, the stored magnetic energy in the system is:

$$W = -\frac{a^2\mu_0 L^2}{16}\sum_{n=1}^{\infty}\sum_{n'=1}^{\infty} j_{\phi n}^a j_{\phi n'}^a \int_{-\infty}^{\infty} dk\,\psi_n(k)\psi_{n'}(k)\, I'_1(ka)K'_1(ka)\left(1-\frac{I'_1(ka)K'_1(kb)}{I'_1(kb)K'_1(ka)}\right) \quad (42),$$

and the expression for the x-directed Lorentz force is written as:

$$F_x^{net} = aL\sum_{n=1}^{\infty} j_{\phi n}^a \int_0^{+\infty}\psi_n(k)\,dk\left\{\left[B_{z0}^{*a}(k)+\frac{1}{ka}B_{\rho 0}^{*a}(k)\right] - \frac{I'_1(ka)}{I'_1(kb)}\left[B_{z0}^{*b}(k)+\frac{1}{kb}B_{\rho 0}^{*b}(k)\right]\right\} \quad (43).$$

The functional E is constructed and has the identical form of Equation (22). However, in the present case, the dependence of E is over $j_{\phi n}{}^a$ and the force is directed along the x-axis instead of the z-axis. Thus, minimizing E with respect to $j_{\phi n}{}^a$ yields a matrix equation for $j_{\phi n}{}^a$ as follows:

$$\sum_{n'=1}^{\infty} j_{\phi n'}^a\left\{\frac{aL\pi}{2}\int_0^{\infty} dk\, I'_1(ka)K'_1(ka)\left(1-\frac{I'_1(ka)K'_1(kb)}{I'_1(kb)K'_1(ka)}\right)\psi_n(k)\psi_{n'}(k)\right\}$$
$$=\sum_{j=1}^{N_1}\lambda_j\cos(\phi_j)\int_0^{\infty} dk\,k\cos kz_j\, I_1(k\rho_j)K'_1(ka)\left(1-\frac{I'_1(ka)K'_1(kb)}{I'_1(kb)K'_1(ka)}\right)$$
$$+\sum_{j=N_1+1}^{N_1+1}\lambda_j\frac{2\pi}{\mu_0}\int_0^{\infty} dk\,\psi_n(k)\left\{\left[B_{z0}^{*a}(k)+\frac{1}{(ka)}B_{\rho 0}^{*a}(k)\right] - \frac{I'_1(ka)}{I'_1(kb)}\left[B_{z0}^{*b}(k)+\frac{1}{kb}B_{\rho 0}^{*b}(k)\right]\right\} \quad (44).$$

Truncating the infinite summations at M terms, and using compact notation, Equation (38) simplifies to:

$$\sum_{n'=1}^{M} j_{\phi n'}^a C_{n'n} = \sum_{j=1}^{N_1+1}\lambda_j D_{jn} \quad (45),$$

or:

$$\underline{J}^a\underline{C}=\underline{\lambda}\ \underline{D} \text{ or } \underline{J}^a=\underline{\lambda}\ \underline{D}\ \underline{C}^{-1} \quad (46),$$

where $\underline{J}^a$ is a 1xM matrix, $\underline{C}$ is a MxM matrix, $\underline{\lambda}$ is a 1xN_1+1 matrix, and $\underline{D}$ is a N_1+1xM matrix with:

$$D_{jn} = \begin{cases} \text{B-field for } j = 1, N_1 \\ F_z^{net} \quad \text{for } j = N_1+1,\ N_1+1 \end{cases} \quad (47).$$

Because Equation (46) is the same as Equation (28), the expressions of the continuous current distribution for both coils is found by following the steps described above. To discretize both current densities, one first considers the continuity equation for the current density:

$$\vec{\nabla}\cdot\vec{J} = 0 \quad (48).$$

In analogy with the magnetic field, where a vector potential is introduced, the current density is expressed as a curl of the stream function $\vec{S}$ as:

$$\vec{J} = \vec{\nabla}\times\vec{S} \quad (49).$$

Because the current is restricted to flow on the surface of a cylinder of radius $a=\rho_a$ and has only angular and axial

dependence, the relation between the current density and the stream function in cylindrical coordinates is:

$$j_\phi^a(\phi,z)\hat{\alpha}_\phi + j_z^a(\phi,z)\hat{\alpha}_z = \frac{\partial S_\rho}{\partial z}\hat{\alpha}_\phi - \frac{1}{a}\frac{\partial S_\rho}{\partial \phi}\hat{\alpha}_z \quad (50),$$

and S_ρ is found from:

$$S_\rho(\phi,z) = -a\int_{-\pi}^{\phi} d\phi' j_z^a(\phi',z) \quad (51).$$

The contour plots of the current density are determined by:

$$S_\rho(x,z) = \left(n - \frac{1}{2}\right) S_{inc} + S_{min} \quad \text{for } n = 1, \ldots, N \quad (52),$$

where N is the number of the current contours, S_{min} is the minimum value of the current density, and S_{inc} represents the amount of the current between two contour lines. The determination of S_{inc} is:

$$S_{inc} = \frac{S_{max} - S_{min}}{N} \quad (53),$$

with S_{max} representing the maximum value of the current density. The contours which are generated by this method follow the flow of the current and the distance between them corresponds to a current equal to an amount of S_{inc} in amps. The discrete wires are positioned to coincide with these contour lines.

In the preferred embodiment, the self-shielded force free x-gradient coil has a radius of the inner coil of about 1/3 metre and a length which is about -100 to 900 mm. The radius of the outer coil is about 0.4 metres. For the design of this self-shielded coil, the number of Fourier coefficients which are used is equal to the total number of constraint points increased by 1. In addition, four constraint points are chosen to specify the quality of the magnetic field inside a 40-50 cm. by 25 cm. ellipsoidal volume, and to eliminate the net Lorentz force for the whole gradient coil system. The first constraint sets the strength of the gradient field, e.g. 13.5 mT/m. The second constraint specifies the linearity of the gradient field along the x-axis and up to the distance of 25 cm. from the isocenter of the gradient field, e.g., 5%. The third constraint specifies the uniformity of the gradient field along the z-axis at a distance up to ±6 cm. from the isocenter of the gradient field, e.g., 12%. The fourth constraint specifies the value of the net x-directed force on the gradient sets location, e.g., 1×10^{-7} Newtons.

The presence of this set of constraints generate a continuous current distribution for both the inner and outer coils which, when discretized, takes the form illustrated in FIGURES 3A and 3B. More specifically, applying these design requirements generates a continuous current distribution of j_{za} and j_{zb}. Only the z-component of the current density is necessary for the creation of a discrete current pattern. The constraints are also chosen such that an integer number of turns for the inner and outer coils with a constant amount of current per loop can be obtained. Using the stream function technique, the discrete current distribution of the inner coil for the lateral force free configuration is obtained as illustrated in FIGURE 3A. It will be noted that the inner or primary coil has a thumbprint winding 70 and an additional or force offsetting winding 72. The current in the force correcting winding effectively flows in an opposite spiral direction to the primary coil portion 70. The current distribution and number of turns in the force cancelling coil portion 72 is selected to zero the net x or y-direction force generated by the primary coil section 70 of FIGURE 3A. Depending on the nature of the B0 field components in a particular design, similar force free results can also be obtained by placing a winding having an opposite sense than the secondary x or y-gradient coil near an outer edge of that coil. The amperage through the force offsetting windings or the number of windings are adjusted until a current density is achieved which produces an x or y-force component which is equal and opposite to the force component generated by the other x or y-gradient coils.

Although force balancing coils are applied to only the primary winding of shielded gradient coils in the preferred embodiment, it is to be appreciated that the primary and secondary coils can each be individually force balanced. Further, it is to be appreciated that the thrust force balancing is also applicable to gradient coils which are not self-shielded, i.e., which have only a primary winding and no secondary shield winding.

Claims

1. A magnetic resonance imaging apparatus which includes a main magnet (10) for generating a temporally constant magnetic field through a patient receiving bore (12) and a gradient field coil assembly (30) and an RF coil (40) around the bore (12) and in which the temporally constant magnetic field has components which interact with current pulses applied to principal windings (32, 36) of the gradient field coil assembly (30) to generate magnetic field gradients in such a manner that a net force is experienced by the gradient field coil assembly (30), characterised

by force correction windings (64, 72) mounted adjacent edge areas of the gradient field coil assembly (30) which carry currents in an opposite sense to the principal windings (32, 36) such that the net force is substantially cancelled.

2. A magnetic resonance imaging apparatus as set forth in claim 1 wherein the patient receiving bore (12) has a length to diameter ratio of less than 1.5:1.

3. A magnetic resonance imaging apparatus as set forth in claim 1 wherein the patient receiving bore (12) has a length to diameter ratio of substantially 1:1.

4. A magnetic resonance imaging apparatus as set forth in any preceding claim wherein the gradient coil assembly (30) is a self-shielded gradient coil having a primary coil (32) and a shield coil (36), current pulses passing through the primary and shield coils (32, 36) generating magnetic fields that combine within the gradient coil assembly (30) to form a linear magnetic field gradient and tending to cancel outside the gradient coil assembly (30), and the current pulses that pass through the shield coil (36) interact with radial magnetic field components to generate a first force, the primary gradient coil (32) being wound such that the current pulses passing through the primary gradient coil (32) interact with the radial magnetic field components to generate a second force that is substantially equal and opposite to the first force.

5. A magnetic resonance imaging apparatus as set forth in any one of Claims 1 to 3 wherein the gradient coil assembly (30) includes primary and secondary gradient coils (32, 36) which each include four symmetrically arranged thumbprint coils (70) for generating a magnetic field gradient transverse to a longitudinal axis of the bore (12), the thumbprint coils (70) further generating a net transverse force, the force correction windings including force offsetting current loops (72) disposed adjacent an end of at least one of the primary and secondary gradient coils (32, 36) for carrying current flows which circulate in an opposite direction to a most adjacent thumbprint coil (70) for offsetting the net transverse force along the transverse direction.

6. A magnetic resonance imaging apparatus as set forth in any one of Claims 1 to 3 wherein the gradient coil assembly (30) includes a primary z-gradient coil (32) including a series of loop coils (60) which are mounted around a cylindrical former (34) for generating a z-gradient field and the force correction windings include loops (64) extending around ends of the cylindrical former (34) and connected in series with the z-gradient loop coils (60) for carrying common current pulses but in an opposite direction.

7. A magnetic resonance imaging apparatus as set forth in any one of Claims 1 to 3 wherein the gradient coil assembly (30) includes a primary z-gradient coil (32) and a secondary, shield z-gradient coil (36) and the secondary z-gradient coil (36) is disposed surrounding the primary z-gradient coil (32), current pulses applied to the primary and secondary z-gradient coils (32, 36) causing magnetic fields which combine within the gradient coil assembly (30) to generate a linear gradient axially along the coil assembly (30) and which generally cancel exterior to the coil assembly (30).

8. A magnetic resonance imaging apparatus as set forth in claim 7 wherein the primary and secondary gradient coils (32, 36) include annular loops (60, 62) and the force correction windings include loops (64) disposed on the ends of at least one of the primary and secondary coils (32, 36).

9. A magnetic resonance imaging apparatus as set forth in any one of Claims 1 to 3 wherein the gradient coil assembly (30) has a z-gradient coil including: a cylindrical former (34); a first plurality of loops (60) extending around the former (34) on a first side of the isocenter of the temporally constant magnetic field, the first plurality of loops (60) carrying current pulses primarily in a first direction circumferentially around the former (34); a second plurality of loops (60) extending around the former (34)on a second side of the isocenter mirror image symmetric about the isocenter with the first plurality of loops (60), the second plurality of loops (60) carrying the current pulses primarily in a second direction circumferentially around the former (34), the second direction being opposite to the first direction, the current pulses passing through the first and second plurality of loops (60) interacting with the radial components of the temporally constant magnetic field to cause a net axial force; first force offset coils (64) extending around the former (34)on the first side of isocenter displaced from the isocenter, the first force offset coils carrying the current pulses in the second direction; and second force offset coils (64) extending around the former (34) on the second side of isocenter and displaced from the isocenter, the second force offset coils (64) carrying the current pulses in a first direction and the current pulses passing through the first and second force offset coils (64) interacting with the radial components of the temporally constant magnetic field to cause an axial force which offsets the net axial force.

10. A method of designing gradient coils for a magnetic resonance imaging apparatus in which coil windings patterns are constrained to have a preselected gradient field strength, at least a preselected gradient field linearity and, at least a preselected gradient field uniformity, characterized by: constraining the coil windings such that the gradient coil assembly experiences substantially no axial or transverse force when current pulses flowing through the gradient coil windings interact with radial and axial components of a main magnetic field within which the gradient coil is disposed.

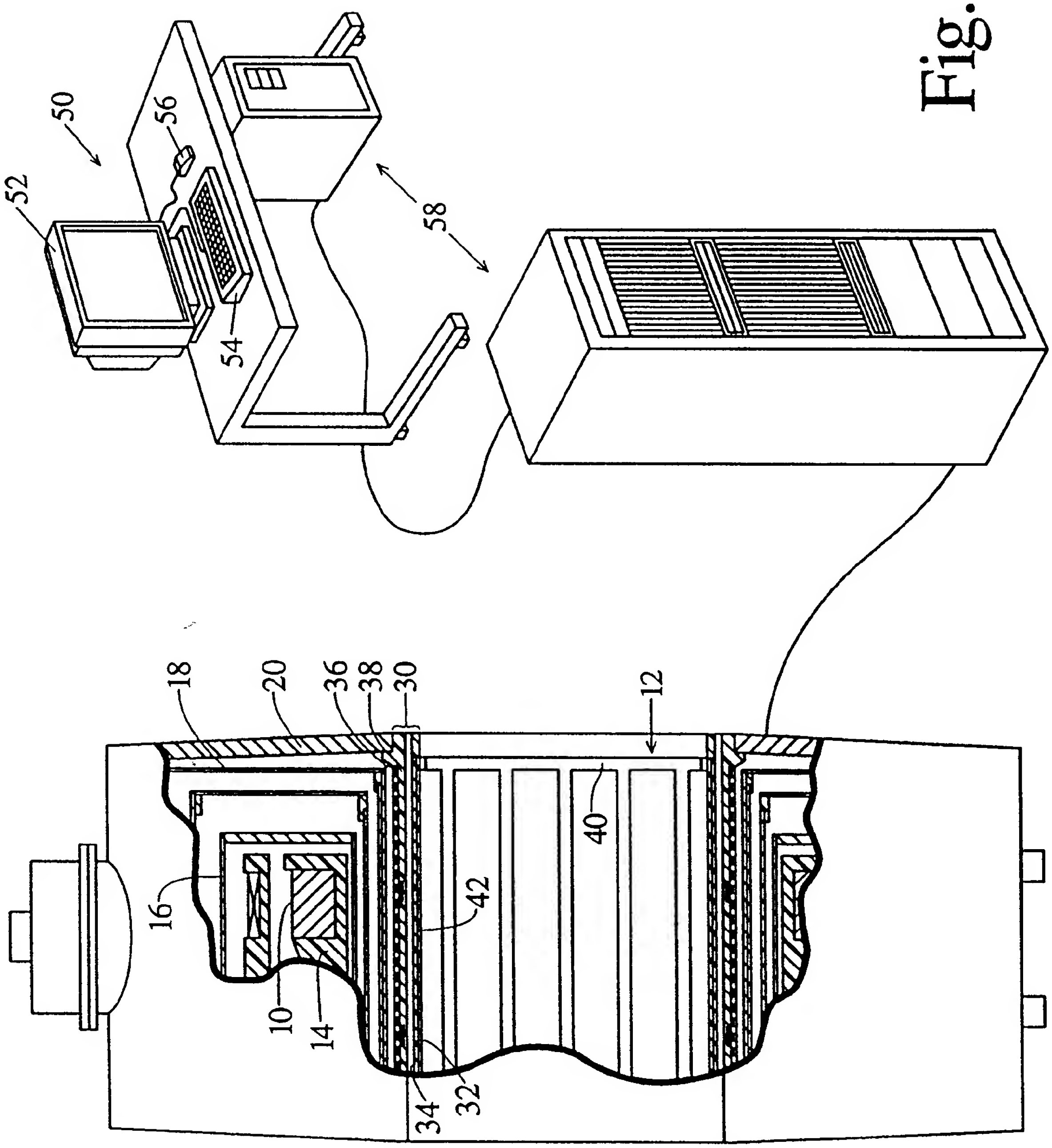

Fig.1

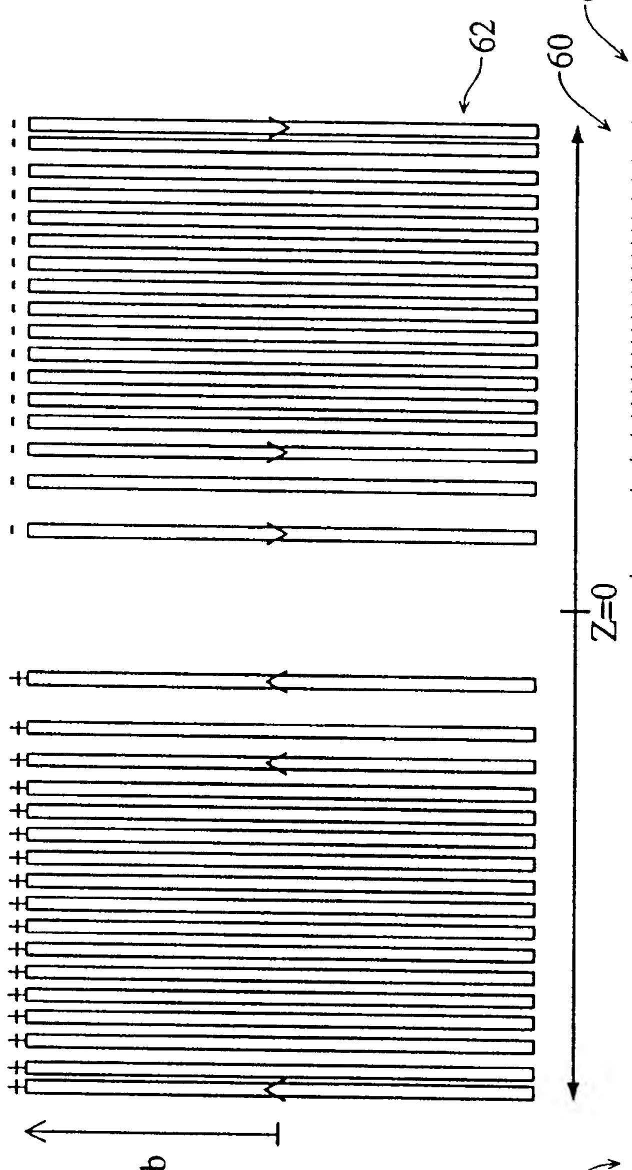

Fig.2B

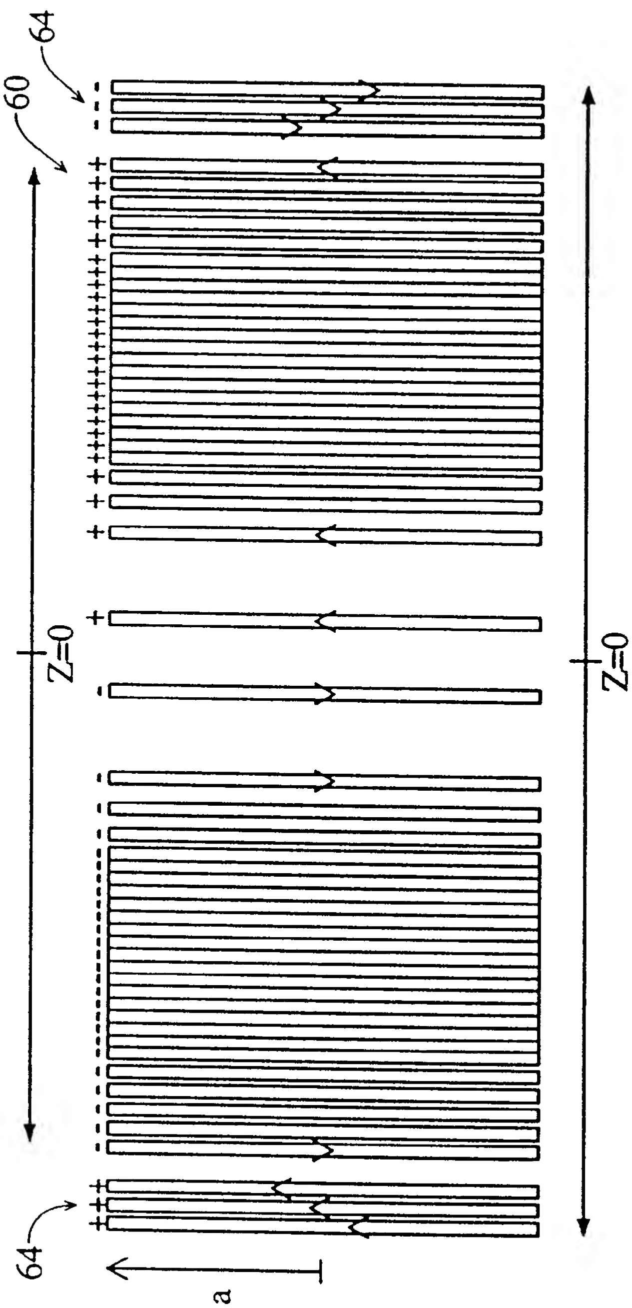

Fig.2A

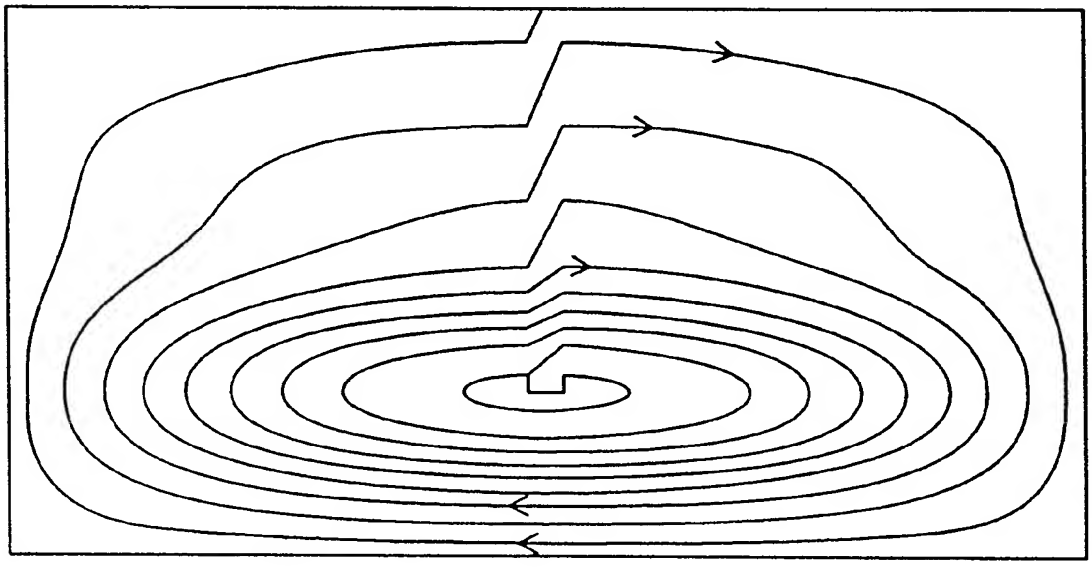

Fig.3B

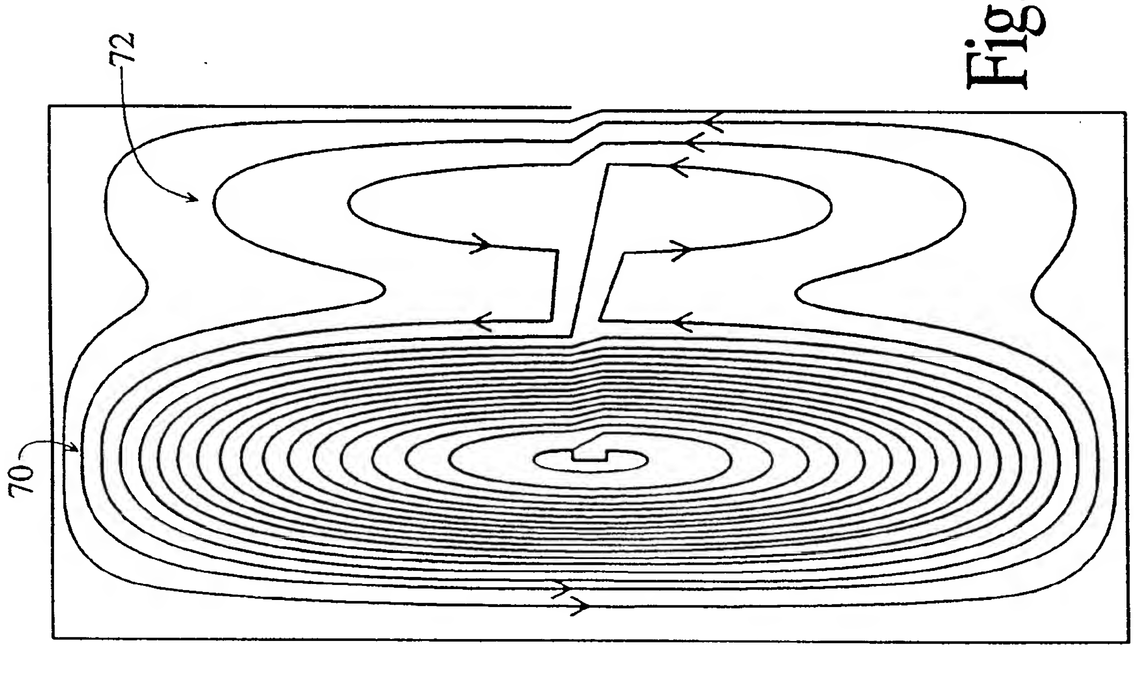

Fig.3A

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EUROPEAN SEARCH REPORT

Application Number
EP 95 30 5393

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.6)
X A	EP-A-0 231 879 (GENERAL ELECTRIC COMPANY) * page 2, line 20 - page 3, line 23 * * page 6, line 3 - page 7, line 27 * * page 10, line 25 - page 11, line 3 * * figures 2,4-6,8,9 * ---	1,4-10 2,3	G01R33/385
X A	US-A-5 198 769 (G. FRESE, E. STETTER) * column 2, line 3 - line 58 * * column 4, line 29 - column 5, line 40 * * figures 4,5,10,11 * ---	1,4,6-8,10 2,3	
X A	MAGNETIC RESONANCE IN MEDICINE, vol. 31, no. 4, April 1994 pages 450-453, A.M. ABDULJALIL ET AL.: 'Torque Free Asymmetric Gradient Coils for Echo Planar Imaging' * the whole article * ---	1,4,6,7,10 5	TECHNICAL FIELDS SEARCHED (Int.Cl.6)
X A	GB-A-2 262 808 (BRUKER ANALYTISCHE MESSTECHNIK) * page 5, line 11 - page 6, line 2 * * page 11, line 16 - page 12, line 18 * * page 20, line 20 - page 21, line 19 * * figure 5 * --- -/--	1,4,10 2,3,6	G01R

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
THE HAGUE	19 January 1996	Volmer, W

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
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EUROPEAN SEARCH REPORT

Application Number
EP 95 30 5393

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.6)
X A	WO-A-93 04493 (BRITISH TECHNOLOGY GROUP USA, INC.) * page 5, paragraph 3 - page 6, paragraph 2 * * page 15, paragraph 2 - page 16, paragraph 2 * * figures 7-13 * ---	1,4-7,10 2,3,9	
P,X	WO-A-94 28430 (BRITISH TECHNOLOGY GROUP LIMITED) * page 4, line 12 - line 29 * * page 5, line 6 - page 7, line 19 * * page 8, line 14 - line 34 * * page 9, line 31 - page 10, line 4 * * page 17, line 19 - page 21, line 19 * * page 22, line 4 - line 28 * -----	1,4,5, 7-9	

TECHNICAL FIELDS SEARCHED (Int.Cl.6)

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
THE HAGUE	19 January 1996	Volmer, W

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CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
Y : particularly relevant if combined with another document of the same category
A : technological background
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